Frontal Plane Standing Balance with an Ambulation Aid: Upper Limb Biomechanics

James Y Tung^{1,4,7}, William H Gage^{2,7}, Karl F Zabjek^{1,3,7}, Brian E Maki^{5,6,7}, William E
McIlroy^{1,4,7}

¹*Graduate Department of Rehabilitation Science, University of Toronto, Toronto, Canada*

²*School of Kinesiology and Health Science, York University, Toronto, Canada* ³*Department of*

Physical Therapy, University of Toronto, Toronto, Canada ⁴*Department of Kinesiology,*

University of Waterloo, Waterloo, Canada ⁵*Centre for Studies in Aging, Sunnybrook Research*

Institute, Toronto, Toronto, Canada ⁶*Department of Surgery, University of Toronto, Toronto,*

Canada ⁷*Toronto Rehabilitation Institute, Toronto, Canada*

Corresponding Author: James Y Tung

Phone: (519) 886-4567 x33007

Email: james.tung@uwaterloo.ca

Word count: 2518

Abstract

Despite widespread acceptance of clinical benefits, empirical evidence to evaluate the advantages and limitations of ambulation aids for balance control is limited. The current study investigates the upper limb biomechanical contributions to the control of frontal plane stability while using a 4-wheeled walker in quiet standing. We hypothesized that: 1) upper limb stabilizing moments would be significant, and 2) would increase under conditions of increased stability demand. Factors influencing upper limb moment generation were also examined. Specifically, the contributions of upper limb center-of-pressure (COP_{hands}), vertical and horizontal loads applied to the assistive device were assessed. The results support a significant mechanical role for the upper limbs, generating 27.1% and 58.8% of overall stabilizing moments under baseline and challenged stability demand conditions, respectively. The increased moment was achieved primarily through the preferential use of phasic upper limb control, reflected by increased COP_{hands} (baseline vs. challenged conditions: 0.29 vs. 0.72 cm). Vertical, but not horizontal, was the primary force direction contributing to stabilizing moments in quiet standing. The key finding that the upper limbs play an important role in effecting frontal plane balance control has important implications for ambulation aid users (e.g., elderly, stroke, traumatic brain injury).

Keywords

Human Balance, Ambulation Aids, Quiet Standing

Introduction

Worldwide, millions of people rely on ambulation aids to address mobility impairments (LaPlante, 1992; Statistics Canada, 2007). The 4-wheeled walker is a device frequently prescribed to facilitate independent standing and walking despite high rates of injury and deaths associated with walker use (Charron et al., 1995). While studies have reported improved walking distance (Solway et al., 2002) and efficiency (Probst et al., 2004) in cardiopulmonary populations, little empirical evidence exists to evaluate 4-wheeled walker use for balance (Bateni and Maki, 2005). Bateni et al. (Bateni et al., 2004) demonstrated that healthy young adults reduced the need for compensatory stepping in response to lateral perturbations when using standard (4-footed) walkers. However, this benefit may be offset by the restrictions to lateral stepping and the potential tripping hazard imposed by the walking frame. Importantly, the role of the upper limbs in whole-body balance control has received little attention compared to the body of research investigating lower limb control.

While the upper limbs can provide tactile input when touch is applied to a stationary object (Jeka, 1997), the mechanical role of the upper limbs in control of standing balance remains to be examined. Cordo & Nashner (1982) and Elger et al. (1999) examined upper limb responses to perturbations while grasping handholds highlighting the potential importance of rapid upper limb reactions to maintain stability. The focus of the present study was to investigate the mechanical contributions of the upper limbs to the control of frontal plane stability in quiet standing with a fixed 4-wheeled walker. Specifically, the objective was to determine if the upper limbs play a significant mechanical role in maintaining standing balance with an ambulation aid and to detail the upper limb strategies and control characteristics.

Considering the inverted pendulum model for standing balance in (unassisted) bipedal standing, frontal plane stability is controlled by modulating the magnitude of the vertical ground reaction force generated by the lower limbs (Winter et al., 1996). The net stabilizing moment is a product of: 1) the moment arm, defined as the horizontal distance between the center of pressure (COP) and center of mass (COM), and 2) the vertical load. In quiet standing where the COM movement is typically small, the moment arm is characterized primarily by the COP displacement (Winter, 1995). With the availability of supports (e.g., a walker), the upper limbs can provide mechanical and sensory influences on balance control. As shown in Figure 1, the upper limbs can contribute to the generation of frontal plane stabilizing moments (about the axis O) by applying: 1) horizontal forces (H_{1x} , H_{2x}) acting through the moment arm defined by the handle height (h); and/or 2) vertical forces (H_{1z} , H_{2z}) acting through the distance defined by the lateral placement of the hands (b). L_{1z} , L_{2z} , and a represent the vertical forces generated by the legs, acting through the moment arm defined by the lateral distance of the feet (a). Overall, the net frontal plane stabilizing moment acting on the body (M_{net}) is defined by:

$$(1) \quad M_{net} = [(L_{1z} - L_{2z})a] + [(H_{1z} - H_{2z})b + (H_{1x} + H_{2x})h]$$

In the current study, the first hypothesis was that the availability of the fixed 4-wheeled walker would lead to substantial upper limb moments, relative to the net whole-body moment, to maintain frontal plane stability. The second hypothesis was that the upper limb contribution would increase as the challenge to balance control increases. An additional objective of the study was to examine the upper limb control characteristics. Moments can be generated by increasing: 1) tonic upper limb loading, reflected by mean % body weight (BW) applied, and/or 2) phasic

control, reflected by upper limb COP displacement amplitudes. Furthermore, phasic upper limb control can be generated through horizontal (H_{1x} , H_{2x}) and/or vertical (H_{1z} , H_{2z}) force directions. The contributions of these factors (i.e., tonic vs. phasic; vertical vs. horizontal) were explored to characterize the balance control strategies employed when using a 4-wheeled walker.

Methods

Participants and Tasks

Eleven (11) healthy, young adults (7 female, 20-35y) without balance impairment provided informed consent to participate in the study, approved by the local ethics committee. Sample size calculations ($\alpha = 0.05$, $\beta = 0.9$, effect size = 1.59) based on data collected from the first 5 participants yielded $n=9$. Two additional participants were collected and reported for the study. Participants stood in 4 task conditions: 1) normal challenge stance (feet pelvis-width apart) without touching the 4-wheeled walker [NC-Hands OFF]; 2) normal challenge stance while holding the 4-wheeled walker [NC-Hands ON]; 3) increased balance challenged (IC) stance (feet together, eyes closed, on 1.9 cm thick foam), without touching the 4-wheeled walker [IC-Hands OFF]; and 4) increased challenge stance while grasping the 4-wheeled walker [IC-Hands ON]. Participants stood with their eyes fixated on a target (5m) with instructions to stand still (NC condition). In Hands OFF conditions, participants stood with elbows flexed 20 degrees and palms down to approximate the Hands ON posture. The 4-wheeled walker handles were adjusted to the height of the radial styloid with arms straight, and the wheels were fixed by locking the brakes. No movement of the wheels was observed throughout the study.

Prior to collection trials, participants performed a 30s trial to acclimate to the task and determine preferred foot position. Initially, the ankles were placed in line with the rear wheel axis and participants could reposition the feet in the sagittal plane during the training trial, if needed. Following the training trial, the participants maintained the preferred foot position and the 60s collection trial was performed. The order of conditions was randomized across participants.

Forceplate measures

Four forceplates, embedded within a wooden platform, measured the ground reaction forces (GRF) underneath the feet and the 4-wheeled walker (Figure 2). Data was acquired at 250 Hz and low-pass filtered (2nd order Butterworth, 10 Hz). Moment and COP displacement magnitudes were measured using root-mean-square (RMS) values of the last 30s to avoid non-stationarities.

Lower limb frontal plane moment (M_{feet}) and center-of-pressure (COP_{feet}) were measured directly from the forceplate under the feet (forceplate 3). Upper limb measures were calculated from forceplates 1, 2, and 4, which was required to capture loads from all four wheels of the walker. Resultant horizontal ($F_{x,\text{hands}}$) and vertical ($F_{z,\text{hands}}$) forces were calculated by scalar sum. Mean upper limb vertical loading ($F_{z,\text{hands}}$) was expressed as a percentage of body weight. M_{hands} was calculated from the measured COP from the 3 upper limb forceplates and correcting for the lateral distances from the origin ($d_x = 0.3375\text{m}$; Eq. 2). The net whole-body moment (M_o) was calculated as the sum of M_{hands} and M_{feet} . The horizontal component of upper limb moment (M_{horiz}) was estimated by multiplying the horizontal force ($F_{x,\text{hands}}$) and handle height (h). The vertical load component (M_{vert}) was calculated as the difference between the M_{hands} and

M_{horiz} . All moments were calculated with respect to the axis of the inverted pendulum (Figure 1, O).

$$(2) \quad M_{\text{hands}} = (\text{COP}_{x1} \cdot F_{z1}) - (d_x + \text{COP}_{x2}) \cdot F_{z2} + (d_x - \text{COP}_{x4}) \cdot F_{z4}$$

$$(3) \quad M_O = M_{\text{hands}} + M_{\text{feet}}$$

$$(4) \quad M_{\text{horiz}} = F_{x,\text{hands}} \times h$$

$$(5) \quad M_{\text{hands}} = M_{\text{horiz}} + M_{\text{vert}}$$

Analysis

The upper limb mechanical contribution was measured using the ratio $M_{\text{hands}}/M_{\text{net}}$, expressed as a percentage of whole-body moment. The hypothesis that upper limb moments are significant was tested using a t-test against the null hypothesis that $M_{\text{hands}}/M_{\text{net}}$ is zero. The second hypothesis that upper limb contributions would increase with additional balance challenge was tested using a paired t-test against the null hypothesis that $M_{\text{hands}}/M_{\text{net}}$ are equal under NC and IC conditions.

To examine the influence of tonic and phasic control of upper limb moments, the task-dependent changes to mean $F_{z,\text{hands}}$ and $\text{COP}_{\text{hands}}$ were evaluated using paired t-tests. To assess the force direction contributions, the difference ($M_{\text{vert}} - M_{\text{horiz}}$) was calculated. A t-test was used to examine the relative overall contributions, and task-dependence was assessed using a paired t-test. To test for normality, t-tests were applied to the rank-transformed data, yielding essentially the same findings. To assess the effect of upper limb availability [Hands OFF/ON] and balance challenge [NC/IC] on lower limb contributions, separate two-way repeated measures ANOVAs

were used with M_{feet} and COP_{feet} as dependent variables. Statistical significance for all tests was defined as $p \leq 0.05$.

Results

Moments

Overall, the hypotheses that the upper limb moments: 1) contribute significantly to frontal plane balance, and 2) were related to the degree of balance challenge were supported. Figure 3 illustrates a representative time series plot of torques from a single participant. Table 1 summarizes the group statistics from all forceplate measures by condition. As shown in Figure 4, the upper limb contributions: 1) were greater than zero ($p < 0.001$), reflected by $M_{\text{hands}}/M_{\text{net}}$ magnitudes (mean $\pm$ SE: $42.91\pm7.09\%$), and 2) increased from $27.05\pm2.63\%$ in the NC condition to $58.76\pm6.96\%$ under IC conditions ($p < 0.001$), as predicted. The effect of upper limb contributions was reflected by a reduction ($p < 0.001$) in M_{feet} observed under Hands ON conditions (85.68 ± 26.20 Nm) compared to Hands OFF (255.76 ± 26.50 Nm). The task-related increase in M_{feet} was significantly smaller when the upper limbs were available ($p = 0.003$).

COP and F_z

Task-dependent changes to upper limb control characteristics are illustrated in Figure 5. While $\text{COP}_{\text{hands}}$ increased under IC conditions (0.72 ± 0.08 vs. 0.29 ± 0.03 cm; $p = 0.002$), mean $F_{z,\text{hands}}$ remained unchanged (3.03 ± 0.76 vs. 2.51 ± 0.52 %BW, $p = 0.523$). In the lower limbs, COP_{feet} decreased with Hands ON ($p < 0.001$) and increased with IC ($p < 0.001$). A significant interaction effect ([Hands OFF/ON] $\times$ [NC/IC]: $p = 0.002$) was observed, supporting the reduced

role of the lower limbs when the upper limbs are available, particularly under challenged conditions.

Upper limb M_{vert} and M_{horiz} contributions

Figure 6 compares upper limb vertical and horizontal force moment contributions. Across conditions, the difference between M_{vert} and M_{horiz} was greater than zero (42.13 ± 7.03 Nm, $p < 0.001$). M_{vert} demonstrated a greater increase in the IC task, reflected by the increase in the difference measure (NC vs. IC ; 19.33 ± 2.21 vs. 64.93 ± 9.93 Nm; $p < 0.001$). Across conditions, mean M_{vert} amplitudes were 2.47 times the mean M_{horiz} magnitude.

Discussion

Overall, the present study supports the role of the upper limbs in generating significant mechanical contributions to frontal plane stability when using an assistive device. The key findings of the study in healthy adults are that: 1) the upper limbs contribute significant frontal plane stabilizing moments, 2) the upper limbs can be preferentially selected to meet demands of the balance task, and 3) moments are generated principally through vertical forces. While previous work has emphasized the upper limb haptic contributions for balance (Jeka, 1997), such studies limited the upper limb mechanical contributions. Participants in the present study were not restricted (or instructed) to use the handles in any particular way. However, all participants used the upper limbs to generate substantial stabilizing moments and adapted to challenging balance tasks by increasing upper limb contributions. While there may be significant haptic contributions to balance control, the current study emphasizes the relevance and characteristics of the mechanical contributions of the upper limbs.

The findings in young, healthy individuals leads to a future focus on individuals with impaired lower limb balance function to determine the relative differences in reliance on upper limbs. Evidence from clinical populations indicates greater reliance on the upper limbs than the healthy population in the current study (2-3 %BW, Table 1). A progressive supranuclear palsy patient with balance impairments recorded 30-35% BW on a 4-footed walker during gait (Fast et al., 1995). Studies investigating the effects of 4 footed walkers instructed healthy participants to simulate reliance by preloading with 20% BW (Bateni et al., 2004; Maki et al., 2008). Even patients with chronic obstructive pulmonary disorder demonstrated a higher loading levels (7% BW) during gait (Solway et al., 2002). Hence, factors that potentially limit upper limb functioning (e.g., arthritis, weakness) may influence the strategies and effectiveness of 4-wheeled walker use for balance.

With respect to strategy, the upper limb contribution was influenced primarily by phasic control in response to (or anticipation of) changes to the COM, indicated by the task-dependent increase in COP_{hands} , but not mean $F_{z,hands}$. The observed interaction (Hands ON x IC) effect on COP_{feet} supports the reduced role of the lower limbs when the upper limbs are available, particularly under challenged conditions. Coupled with the lack of changes to $F_{z,feet}$, the reduced lower limb moment contributions arise from a reduction in phasic COP_{feet} amplitudes. Considering the relatively low levels of mean $F_{z,hands}$ observed in the healthy population sampled, the preference for phasic control strategy over tonic loading of the upper limbs in clinical populations remains to be examined.

Analysis of the force direction contributions revealed that vertical loading is the primary upper limb contribution, averaging 2.47 times greater than the horizontal contributions. Furthermore, M_{vert} amplitude increased from 2.0 times the magnitude of M_{horiz} in the baseline (NC) condition to 2.7 times in the challenged (IC) condition. Considering the relatively low vertical load values (2-3% BW) in healthy adults from the current study, the M_{vert} contribution may be greater in impaired populations that have reported higher levels of vertical load. Since the horizontal direction is applied through a longer moment arm (Figure 1, h) compared to vertical (b), horizontally applied force is mechanically advantageous. However, excessive horizontal forces have been associated with the risk of tipping walking frames (Pardo et al., 1993; Finkel et al., 1997). The results from the current study support a stationary standing strategy that prioritizes assistive device stability over mechanical efficiency.

The main limitation of the study is the lack of data from populations who use ambulation aids regularly for balance and mobility, relying on simulation of challenged balance in young, healthy adults. There is a need to verify the findings from the current study to populations with balance impairments. Furthermore, the current study examined the overall mechanical contribution of the upper limbs with the 4-wheeled walker for balance assistance, including the benefit of sensory influences from the upper limbs. The lack of means to separate the haptic influences from the overall effects limits the specific examination of mechanical contributions.

This study was conducted within the context of on-going efforts to evaluate the effectiveness of 4-wheeled walker assistive devices for balance and mobility in everyday activities (Tung, JY et al., 2007). While the contribution of the horizontal forces are not negligible, the vertical moment

contributions were nearly 2.5 times the horizontal force contributions supporting the potential of vertical-load measures to indicate upper limb involvement . Specifically, the vertical forces measured from uniaxial load cells mounted vertically in the walker legs may be a useful proxy measure of upper limb involvement for frontal plane balance control. On-going studies include continuing to investigate of the role of the upper limbs during dynamic, walking tasks that may provide an important means to assess safe device use, treatment progress, and prescription criteria.

Conflict of Interest Statement

The authors have no conflicts of interest to declare.

Acknowledgements

We acknowledge the support from Canadian Institutes of Health Research, Natural Sciences and Engineering Research Council of Canada, Toronto Rehabilitation Institute who receives funding under the Provincial Rehabilitation Research Program from the Ministry of Health and Long Term Care in Ontario, and the National Institute on Disability and Rehabilitation Research (NIDRR) through the Rehabilitation Engineering Research Center on Universal Design and the Built Environment (grant #H133E050004-08A), a partnership with the Centre for Inclusive Design and Environmental Access (IDEA).

References

Batani, H., Heung, E., Zettel, J., McIlroy, W.E., Maki, B.E., 2004. Can use of walkers or canes impede lateral compensatory stepping movements? *Gait & Posture* 20, 74-83.

- Bateni, H., Maki, B.E., 2005. Assistive devices for balance and mobility: benefits, demands, and adverse consequences. *Archives of Physical Medicine & Rehabilitation* 86, 134-45.
- Charron, P.M., Kirby, R.L., MacLeod, D.A., 1995. Epidemiology of walker-related injuries and deaths in the United States. *American Journal of Physical Medicine & Rehabilitation* 74, 237-239.
- Cordo, P.J., Nashner, L.M., 1982. Properties of postural adjustments associated with rapid arm movements. *Journal of Neurophysiology* 47, 287-302.
- Elger, K., Wing, A., Gilles, M., 1999. Integration of the hand in postural reactions to sustained sideways force at the pelvis. *Experimental Brain Research* 128, 52-60.
- Fast, A., Wang, F.S., Adrezn, R.S., Cordaro, M.A., Ramis, J., Sosner, J., 1995. The instrumented walker: Usage patterns and forces. *Archives of Physical Medicine and Rehabilitation* 76, 484-491.
- Finkel, J., Fernie, G., Cleghorn, W., 1997. A guideline for the design of a four-wheeled walker. *Assistive Technology* 9, 116-129.
- Jeka, J.J., 1997. Light touch contact as a balance aid. *Physical Therapy* 77, 476-87.
- LaPlante, M.P., 1992. Assistive technology devices and home accessibility features: prevalence, payment, need, and trends. *Advance Data from US Vital and Health Statistics* 13.
- Maki, B.E., Cheng, K.C., Mansfield, A., Scovil, C.Y., Perry, S.D., Peters, A.L., McKay, S., Lee, T., Marquis, A., Corbeil, P., Fernie, G.R., Liu, B., McIlroy, W.E., 2008. Preventing falls in older adults: New interventions to promote more effective change-in-support balance reactions. *Journal of Electromyography and Kinesiology* 18, 243-254.
- Pardo, R.D., Deathe, A.B., Winter, D.A., 1993. Walker user risk index. A method for quantifying stability in walker users. *American Journal of Physical Medicine & Rehabilitation* 72, 301-5.
- Probst, V.S., Troosters, T., Coosemans, I., Spruit, M.A., Pitta, F.D.O., Decramer, M., Gosselink, R., 2004. Mechanisms of improvement in exercise capacity using a rollator in patients with COPD. *Chest* 126, 1102-7.
- Solway, S., Brooks, D., Lau, L., Goldstein, R., 2002. The Short-term Effect of a Rollator on Functional Exercise Capacity Among Individuals With Severe COPD. *Chest* 122, 56-65.
- Statistics Canada, 2007. The 2006 Participation and Activity Limitation Survey: Disability in Canada (Census statistics No. 89-628-XIE). Federal Government of Canada.
- Tung, J.Y., Gage, W.H., Zabjek, K.F., Brooks, D., Mihailidis, A., Fernie, G.R., Maki, B.E., McIlroy, W.E., 2007. iWalker: A 'Real-World' Mobility Assessment Tool, In *Proceedings of the 30th Canadian Medical and Biology Engineering Conference*, Toronto, Canada.
- Winter, D.A., Prince, F., Frank, J.S., Powell, C., Zabjek, K.F., 1996. Unified theory regarding A/P and M/L balance in quiet stance. *Journal of Neurophysiology* 75, 2334-43.
- Winter, D.A., 1995. Human balance and posture control during standing and walking. *Gait & Posture* 3, 193-214.

Table 1. Group means (SE) of forceplate measures by condition. NC = normal challenge; IC = increased challenge conditions. Notes: upper limb statistics from t-tests; lower limb statistics from 2-way repeated measures ANOVA.

Measure	Hands OFF		Hands ON		Notes
	NC	IC	NC	IC	
Overall					
RMS M _{net} [Nm]	166.31 (24.94)	345.20 (28.06)	82.75 (17.54)	148.98 (29.27)	
Upper limbs					
RMS M _{hands} [Nm]			28.09 (3.07)	73.72 (7.94)	
Upper limb contribution [%]			27.05 (2.63)	58.76 (6.96)	$\mu_0 \neq 0$, $p < 0.001$ $\mu_{\text{NC}} \neq \mu_{\text{IC}}$, $p < 0.001$
RMS COP _{hands} [cm]			0.29 (0.03)	0.72 (0.08)	$\mu_{\text{NC}} \neq \mu_{\text{IC}}$, $p < 0.001$
F _{z, hands} [%BW]			2.51 (0.52)	3.03 (0.76)	$\mu_{\text{NC}} = \mu_{\text{IC}}$, $p = 0.198$
RMS M _{horiz} [Nm]			17.90 (2.83)	37.23 (3.96)	
RMS M _{vert} [Nm]			35.73 (4.64)	100.67 (13.10)	
RMS M _{vert} - RMS M _{horiz} [Nm]			19.33 (2.21)	64.93 (9.93)	$\mu_0 \neq 0$, $p < 0.001$ $\mu_{\text{NC}} \neq \mu_{\text{IC}}$, $p < 0.001$
Lower limbs					
RMS M _{feet} [Nm]	166.31 (24.94)	345.20 (28.06)	80.13 (17.51)	91.22 (34.89)	[Hands OFF/ON]: $p < 0.001$ [NC/IC]: $p < 0.001$ [NC/IC] x [Hands OFF/ON]: $p = 0.003$
RMS COP _{feet} [cm]	0.23 (0.03)	0.55 (0.04)	0.10 (0.02)	0.11 (0.03)	[Hands OFF/ON]: $p < 0.001$ [NC/IC]: $p < 0.001$ [NC/IC] x [Hands OFF/ON]: $p = 0.002$

Figure 1. Free body diagram of the user (left) and the device (right) indicating forces involved in maintaining stability about the axis of rotation (O) during quiet standing. H_{x1} and H_{x2} represent horizontal forces acting through the moment arm defined by the height of the handles (h). H_{z1} and H_{z2} represent vertical forces through the moment arm defined by the width of the handles (b). Upward (i.e., pulling) forces to the handles are considered as negative vertical forces.

Figure 2. Experimental setup using 4 forceplates. Forceplates 1, 2, and 4 were placed under the 4-wheeled walker. Forceplate 3 was used to collect lower limb measures.

Figure 3. Representative time series plots of M_{hands} (upper panel) and M_{feet} (lower) for single participant. Dotted and solid lines indicate Hands ON-NC and Hands ON-IC conditions, respectively.

Figure 4. Mean upper limb moment contributions ($M_{\text{hands}}/M_{\text{net}}$), as a percentage of the net whole-body moment magnitudes. *mean significantly different from zero ($p < 0.001$). #significant difference ($p < 0.001$). Error bars indicate standard error.

Figure 5. Mean upper limb moment M_{hands} (left), $\text{COP}_{\text{hands}}$ (middle), and mean $F_{z,\text{hands}}$ (right) by task condition. *significant difference ($p \leq 0.05$). Error bars indicate standard error.

Figure 6. Mean difference between moments generated through vertical loading (M_{vert}) and horizontal (M_{horiz}) force directions. *significantly different from zero ($p \leq 0.001$). #significant difference ($p \leq 0.001$). Error bars indicate standard error.